3D Finite Element Model of Meniscectomy: Changes in Joint Contact Behavior

Barbara Zielinska

Tammy L. Haut Donahue[1]
Phone: (906) 487-2078
Fax: (906) 487-2822
e-mail: thdonahu@mtu.edu

Department of Mechanical Engineering-Engineering Mechanics,
Michigan Technological University,
1400 Townsend Dr.,
Houghton, MI 49931

The goal of this study is to quantify changes in knee joint contact behavior following varying degrees of the medial partial meniscectomy. A previously validated 3D finite element model was used to simulate 11 different meniscectomies. The accompanying changes in the contact pressure on the superior surface of the menisci and tibial plateau were quantified as was the axial strain in the menisci and articular cartilage. The percentage of medial meniscus removed was linearly correlated with maximum contact pressure, mean contact pressure, and contact area. The lateral hemi-joint was minimally affected by the simulated medial meniscectomies. The location of maximum strain and location of maximum contact pressure did not change with varying degrees of partial medial meniscectomy. When 60% of the medial meniscus was removed, contact pressures increased 65% on the remaining medial meniscus and 55% on the medial tibial plateau. These data will be helpful for assessing potential complications with the surgical treatment of meniscal tears. Additionally, these data provide insight into the role of mechanical loading in the etiology of post-meniscectomy osteoarthritis.
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Introduction

Injuries to the menisci are common due to their high load bearing role in the knee joint. These injuries occur due to traumatic activities as seen in sports, and in older individuals because of age-related material fatigue [1–3]. Until 1948 most meniscal injuries were treated with total meniscectomies. Shortly after 1948, when Fairbank [4] recognized damage to the underlying articular cartilage following total meniscectomy, clinicians began preserving as much meniscus as possible [5]. Today the most common surgical treatments for meniscal injuries include meniscal repair, allograft replacement, and partial meniscectomy [5–8].

In a landmark study by King in 1936 [9], it was shown that lesions in menisci need to communicate with a peripheral blood supply in order to heal. Clinically, injured menisci are examined for potential repair by being classified according to the location of the tear relative to the blood supply, and the "vascular appearance" of the peripheral and central surfaces of the tear [8]. Since the repair of tears in the avascular zone is unlikely to be successful, allografts and meniscectomies are performed on these types of injuries. Difficulty in sizing and fixating meniscal allografts has resulted in limited use of this treatment [10–12]. In fact, improper sizing of meniscal allografts has been shown to decrease the tibial plateau contact area by 23% and increase tibial plateau pressures by 36% [10]. Similarly, meniscectomies decrease the tibial plateau contact area by approximately 10% and increase tibial plateau pressures by approximately 65% [13]. Many studies have shown that partial meniscectomy can lead to destruction of articular cartilage and cause osteoarthritis (OA) [13–19].

Previous experimental and computational studies of partial meniscectomy have focused on meniscectomy induced changes in the underlying articular cartilage. Little attention has been paid to how the remaining meniscal tissue responds following a partial meniscectomy. Following anterior cruciate transection, degenerative changes in the meniscus occur prior to changes in articular cartilage [20]. This raises the possibility that degenerative changes in the meniscus are involved in the early stages of OA development. In fact, following medial partial meniscectomy in rabbits, the menisci expressed an increased amount of nitric oxide (NO), a free radical implicated in the development of OA [21]. A better understanding of the loading environment on the remaining meniscal tissue following meniscectomy may help to better understand the etiology of OA.

In 1985, Aspen et al. [22] developed a simple two-dimensional (2D) axially symmetric finite element (FE) model of the meniscus and showed that the morphology of meniscal tears is dependent on anatomical location. More recently, axisymmetric FE models [23,24] and three-dimensional (3D) FE models [25] showed that meniscal geometry affects the tibio-femoral contact pressures. Haut Donahue [25–27], using FE modeling, studied the importance of meniscal attachments, geometry and material properties in determining the tibio-femoral contact distribution in the human knee joint and found that the contact distribution was sensitive to the stiffness of the horn attachments, as well as the circumferential and axial modulus of the meniscus. Additionally, these data showed that the contact distribution is highly sensitive to both transverse and cross-sectional meniscal geometrical parameters. Brown et al. [28] presented results of contact pressure and contact area on femoral condyles with various angles of flexion for an intact knee, and following total meniscectomy at 0 degrees of flexion, and found a 15% increase in mean contact pressure following a total meniscectomy. Previous studies have documented how partial and total meniscectomy change contact pressure distribution on the tibial plateau [13,29–33]. However, these studies did not quantify how much meniscal tissue was removed or vary the degree of partial meniscectomy. No studies have documented how partial meniscectomy affects the pressure distribution on the remaining meniscal tissue.

The aim of this study was to utilize an existing FE model to study the changes in contact pressure and area on the superior medial and lateral menisci following varying degrees of partial medial meniscectomy. Additionally, articular cartilage contact pressure and area, strain in the medial meniscus and articular cartilage, and the amount of load transmitted through both menisci were quantified. We hypothesized that increasing the percentage of the meniscus that is removed decreases the contact area and

[1]Corresponding author.
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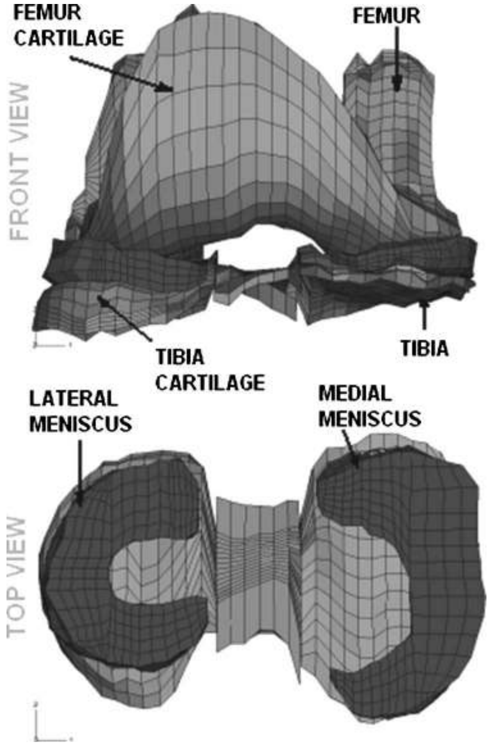

Fig. 1 FE model of a human knee

increases the contact pressure, and that changes in contact behavior are dependent on the site of meniscectomy (anterior, central, posterior).

Method

The FE model (Fig. 1), previously created by Haut Donahue et al. [25,26] was adapted in this study to simulate varying degrees of partial meniscectomy. Briefly, the model was generated from a 3D laser coordinate digitizing system that imaged the cartilage and menisci of one human knee, with the error of less than 8 μm [26]. The solid model was then meshed using commercially available software (TruGrid, XYZ Scientific, Livermore, CA), and contains 8050 hexahedral continuum elements, 1421 contact elements and 13679 nodes. The characteristic length of an element is 2.31 mm. A thorough convergence study was previously conducted on the validated FE model [26]. The convergence results indicated small ($<2\%$) changes in contact variables when decreasing the mesh size to 1 mm$\times$1 mm versus 2 mm$\times$2 mm. Thus, the average element size of 2 mm$\times$2 mm was deemed sufficiently fine and was used for the remainder of the model studies. The FE model includes the femur and tibia modeled as rigid bodies since a previous study confirmed that this simplification had no substantive effect on the contact variables [26]. Menisci (medial and lateral) were treated as linearly elastic and transversely isotropic, while the articular cartilage was considered as a linearly elastic and isotropic material (Table 1). The model also includes meniscal horn attachments, the anterior and posterior cruciate ligaments, the transverse ligament, and the deep medial and lateral collateral ligaments (Table 1). The anterior and posterior cruciate, and deep medial and lateral collateral ligaments were modeled as one-dimensional nonlinear springs, requiring a nonlinear stiffness parameter (k), and a reference strain (ε_r), where reference strain is the initial strain in the reference position (i.e., full extension) [34–37]. All of these ligaments were modeled with anterior and posterior bundles. The transverse ligament and meniscal horn attachments were modeled as linear springs [26], due to lack of data on material behavior in the literature. There were 10 nodes distributed over the part of each meniscus that was considered to be the horn (i.e., where the menisci are attached to the plateau). The area of the transverse ligament was small enough that only one node was in the vicinity of the attachment site. Furthermore, the deep medial collateral ligament attached from the tibia to the medial meniscus, and then from the medial meniscus to the femur. Material properties and characteristics of all elements of the model are shown in the Table 1. Considering that the loading time of interest corresponds to that of a single leg stance, and that the viscoelastic time constant for cartilage approaches 1500 s, the cartilage was assumed to behave as a linearly elastic, isotropic and homogeneous material for the purpose of analyzing contact stresses [38,39]. The elastic modulus (E=15 MPa) and Poisson ratio (ν=0.475) values were selected based on direct measurements under short loading times [40]. The model has been previously validated against experimental measurements of the contact distribution [25]. For loads ranging from 400 to 1200 N and flexion angles of 0 and 15 deg, the model matched experimental measures of contact area, maximum pressure, mean pressure, and location of maximum pressure within 5.4%.

To maintain the same boundary conditions as the previous study, a compressive load of 1200 N was applied to the distal tibia, at 0 deg of flexion. The application point was defined using the coordinate system of Grood and Suntay [41], and the direction of the load was chosen according to the procedures described by Bach and Hull [42]. 1200 N represents 2 times the body weight. The proximal surface of the femur was fixed in space, and rotation about the flexion/extension axis of the tibia was fixed. Contact was modeled between the femur and meniscus, the meniscus and tibia, and the femur and tibia for both the lateral and medial hemi joints, resulting in six contact surface pairs. The contact conditions in the model were completely general involving finite sliding of pairs of curved, deformable surfaces. All surfaces were modeled as frictionless. The contact pressure-clearance relationship used to define the surface interaction was a “hard” contact model, in the sense that nodes were not allowed to penetrate into another surface and no transfer of tensile stress was allowed across the interface [26].

Since the medial meniscus is injured and repaired more often than the lateral meniscus, varying degrees of partial meniscectomy were only examined in the medial meniscus [1]. Simulating partial meniscectomy in the FE model was accomplished by removing elements. To correlate the contact pressures on the superior surface of the meniscus to the percentage of meniscus removed, eleven full-thickness meniscectomy simulations were performed. Varying amounts of tissue were removed from different places in white-white and white-red zones, which correspond to zones with little or no vascularization. The amount of tissue removed was quantified relative to the superior surface of an intact medial meniscus (i.e., superior surface of an intact medial meniscus equals 100%). Since this study is primarily focused on contact behavior, it is more appropriate to quantify percentage of removed medial meniscus based on contact surface area versus meniscal volume. Simulated meniscectomies were chosen to mimic clinical surgeries (Fig. 2). The removal of tissue was limited by a discretization of the tissue into elements. Partial meniscectomies were matched closely to clinical practice and elements were slightly modified at the borders of tissue removal to smooth the edges. The amount of meniscal tissue to remove was calculated using commercially available graphical software (Unigraphics (UG), Unigraphics Solution Inc., Cypress, CA). All nodes from the edge of the superior surface of the medial meniscus were exported to UG as points. A closed curve consisting of straight lines drawn through the meniscal edge points was created and the area was measured. The FE analyses were carried out using commercially available software (ABAQUS 6.4, HKS Inc., Pawtucket, RI).

Mean and maximum contact pressures and contact areas of the superior surfaces of the medial and lateral menisci and the tibial

Table 1 Material parameters for model components

Femoral/tibial cartilage	Linearly elastic, isotropic	$E=15$ MPa, $\nu=0.475$
Lateral/medial menisci	Linearly elastic, transversely isotropic	$E_{\text{axial/radial}}=20$ MPa, $E_{\text{circum}}=150$ MPa, $\nu_{\text{in-plane}}=0.2$, $\nu_{\text{out-of-plane}}=0.3$, Shear modulus=57.7 MPa
Anterior cruciate ligament (ACL)	1D nonlinear spring	Anterior bundle: Reference strain=0.06 Nonlinear stiffness=5000 N Posterior bundle: Reference strain=0.10 Nonlinear stiffness=5000 N
Medial collateral ligament (MCL)	1D nonlinear spring	Anterior bundle: Reference strain=0.0 Nonlinear stiffness=4000 N Posterior bundle: Reference strain=0.0 Nonlinear stiffness=4000 N
Posterior cruciate ligament (PCL)	1D nonlinear spring	Anterior bundle: Reference strain=−0.24 Nonlinear stiffness=9000 N Posterior bundle: Reference strain=−0.03 Nonlinear stiffness=9000 N
Lateral collateral ligament (LCL)	1D nonlinear spring	Anterior bundle: Reference strain=−0.25 Nonlinear stiffness=2000 N Superior bundle: Reference strain=−0.05 Nonlinear stiffness=2000 N Posterior bundle: Reference strain=0.08 Nonlinear stiffness=2000 N
Transverse ligament (TL)	1D linear spring	Stiffness=900 N/mm
Horn attachments	1D linear spring	Stiffness=2000 N/mm

plateau were determined for all simulations. The measurement of contact area is based on the nodes which are in contact, and the associated surfaces of the elements to which these nodes belong. For elements in which all nodes were not in contact on the surface, each node of the element was assigned a portion of the element surface which was included in determining contact area (ABAQUS USER MANUAL, HKS Inc., Pawtucket, RI). The axial strain was computed over the entire surface of the meniscus corresponding to nodal locations on the superior surface. To determine if different regions of the meniscus respond differentially to meniscectomy, the maximum contact pressure was analyzed separately for the anterior, central, and posterior part of the superior medial meniscus surface. To facilitate this, the meniscus was divided into three regions based on the length of the superior medial meniscus surface outer edge (Fig. 2, intact case). To further determine how load transmission is affected by medial meniscectomy, contact force was reported separately for contact between the femur and superior meniscal surfaces and between the articular cartilage of the femur and tibia. Simple linear regression analysis (Minitab Inc., State College, PA) was used to correlate the mean contact pressure, maximum contact pressure, and contact area with the percentage of meniscus removed.

Results

For the intact medial meniscus, contact pressure was greatest in the posterior region with lower pressures in the anterior and central regions. A more uniform distribution of pressure occurred on the lateral meniscus (Fig. 3-black bold line). Removing various portions of the medial meniscus generally did not alter the location of maximum pressure, but increased the magnitude of contract pressure (Fig. 4). The location of the maximum pressure on the medial meniscus was in the outer 1/3 posterior portion of the tissue. Following different meniscectomies the location of maximum pressure moved 3 mm, at most, to an adjacent node. A significant linear correlation was found between the percentage of removed tissue and maximal contact pressure ($p=0.0001$, Fig. 4(A)). The maximum contact pressure on the medial meniscus increased from 4.7 MPa for the intact case to 7 MPa when 60% of the medial meniscus was removed. Strong correlations between the maximum contact pressure and percentage of medial meniscus removed were found for the anterior-central ($p=0.0011$) and posterior-central ($p=0.0093$) groups (Figs. 4(B) and 4(C)). The superior surface of the lateral meniscus had a maximum pressure of 3.6 MPa for the intact medial meniscus. Increases in maximum

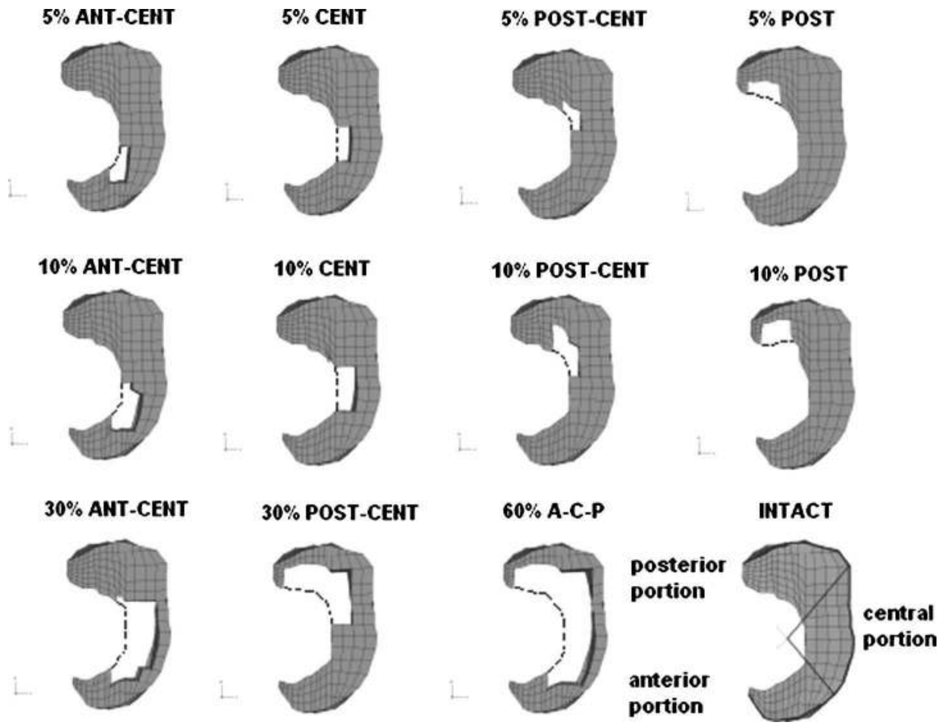

Fig. 2 Partial medial meniscectomies simulated in the model

contact pressure with medial meniscectomy ranged from 6% for 30% anterior-central meniscectomy to 25% for 60% meniscectomy. The maximum contact pressure on the tibial plateau occurred on the lateral side for intact, 5% and 10% meniscectomies (except 10% posterior meniscectomy) and then switched to the medial side of the joint for 30% and 60% meniscectomy. For the intact meniscus, the maximum contact pressure on the tibial plateau was 3.8 MPa. The maximum pressure showed small increases ($\leqslant 10\%$) with 5% and 10% meniscectomies, whereas increases of up to 55% were seen when 60% of the meniscus was removed, resulting in pressures as large as 5.9 MPa.

The mean contact pressure for the superior surface of the medial meniscus increased from 1.57 MPa for the intact case to 3.09 MPa for the 60% meniscectomy showing a similar linear trend ($R^2=0.83$, $p=0.0001$) as maximum contact pressure. The anterior-central group had a slightly stronger correlation between mean contact pressure and percentage of medial meniscus removed ($p=0.0001$) than the posterior-central group ($p=0.0014$). The anterior-central and central meniscectomy caused larger increases in mean pressure on the superior side of the medial meniscus compared to meniscectomies in the posterior region. The

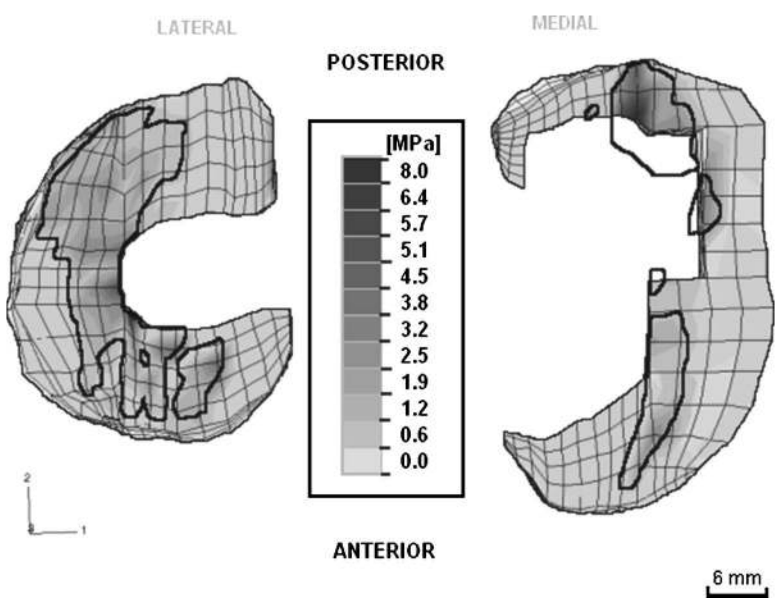

Fig. 3 Contour plot of contact pressures on the superior surface of the lateral and medial meniscus following 30% posterior-central meniscectomy. Black bold line represents border of the contact area for intact case.

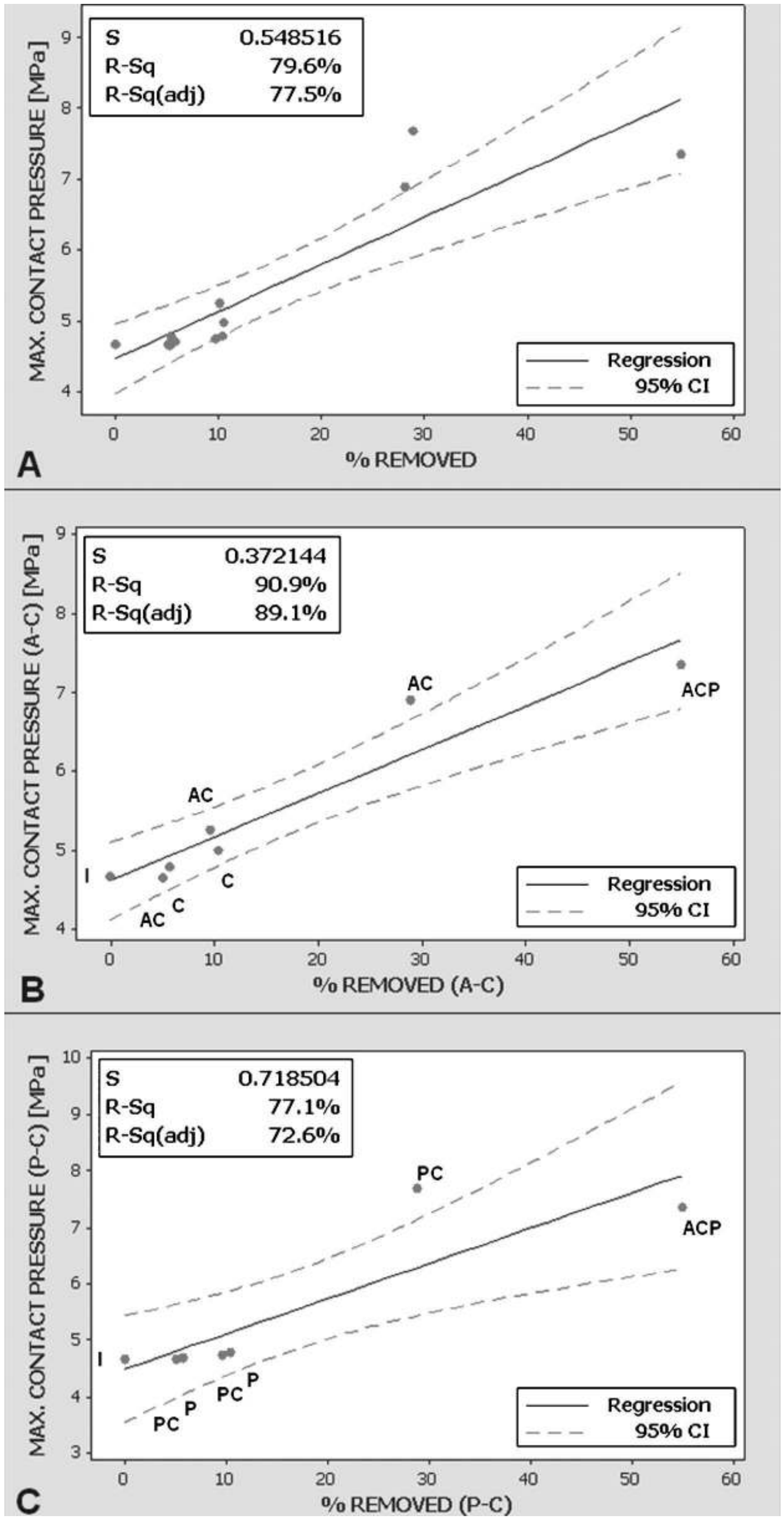

Fig. 4 Linear correlation between the maximum contact pressure and the percentage of removed tissue. (*A*) All experiments, (*B*) anterior-central group, (*C*) posterior-central group. A-anterior, C-central, P-posterior, I-intact.

lateral meniscus showed small changes in mean pressure ($\leqslant 7\%$) following all simulations. Mean contact pressure on the tibial plateau for the intact case was 1.14 MPa and experienced relatively small changes following medial meniscus meniscectomy, reaching the highest increase (26%) following 60% meniscectomy.

Contact area significantly ($p \leqslant 0.0027$) decreased with increased percentage of meniscus removed for the anterior-central and posterior-central groups (Figs. 5(*A*) and 5(*B*)). Changes in contact area on the superior surface of the lateral meniscus were small for all meniscectomies, with a maximum change of only 7%. Following 5% and 10% meniscectomies, up to an 11% de-

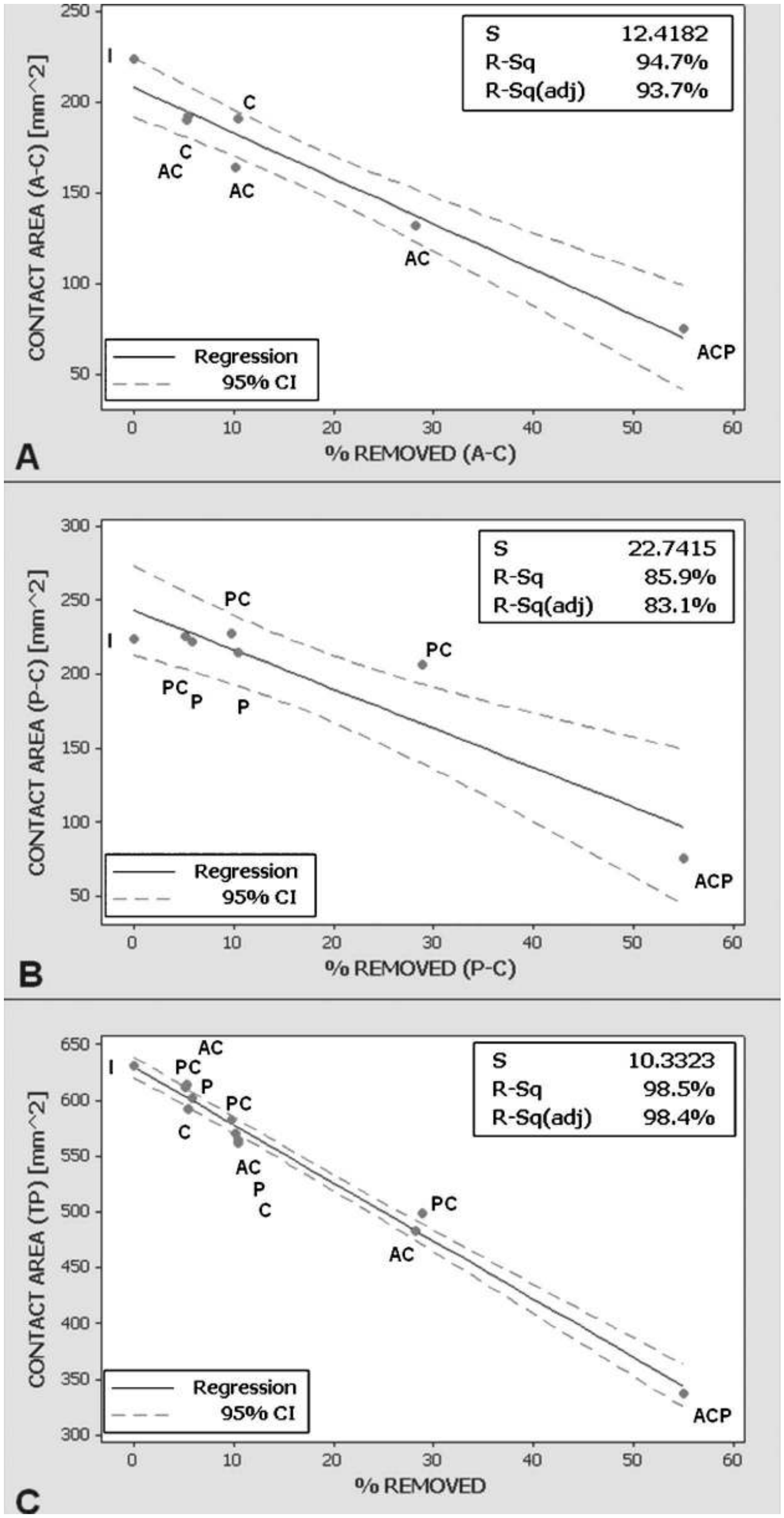

Fig. 5 Linear correlation between the contact area and the percentage of removed tissue. (*A*) Anterior-central group-superior surface of the medial meniscus, (*B*) posterior-central-superior surface of the medial meniscus group, (*C*) Medial hemijoint of the tibia plateau. A-anterior, C-central, P-posterior, I-intact.

crease in the medial tibial plateau contact area was noted. After 60% meniscectomy the contact area decreased by as much as 46% (Fig. 5(*C*)).

The central portion of the superior surface of the intact medial meniscus experienced the smallest maximal contact pressure, and the posterior portion experienced the highest (Figs. 6(*A*)–6(*C*)). The pressure on the posterior portion of the medial meniscus was only affected when 30% and 60% of the medial meniscus was

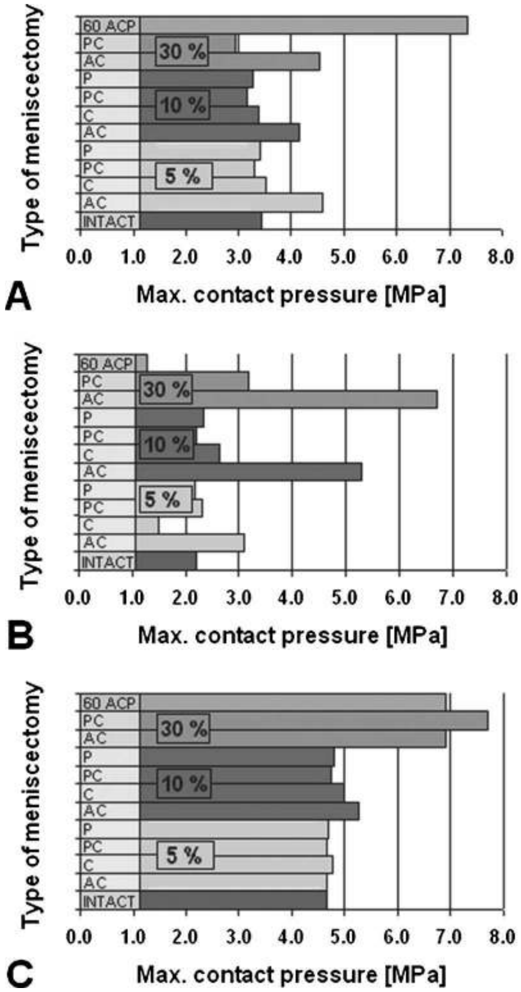

Fig. 6 Maximum contact pressure on the superior surface of the medial meniscus. (*A*) Anterior portion, (*B*) central portion, (*C*) posterior portion.

removed. The anterior and central portions of the medial meniscus were most affected when meniscectomies occur in these regions: 5% anterior-central, 10% anterior-central, 30% anterior-central, and 60% (Figs. 6(*A*) and 6(*B*)).

Due to the limited changes in the contact behavior on the lateral meniscus during the simulated medial meniscus meniscectomies, axial strain analysis was done only for the medial meniscus and articular cartilage on the tibial plateau. A 5% and 10% meniscectomy resulted in small changes in medial meniscus strain (up to 9%, Fig. 7) whereas 30% and 60% meniscectomy almost doubled

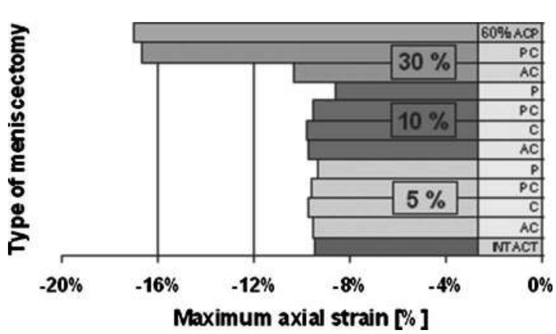

Fig. 7 Maximum axial strain on the superior surface of the medial meniscus

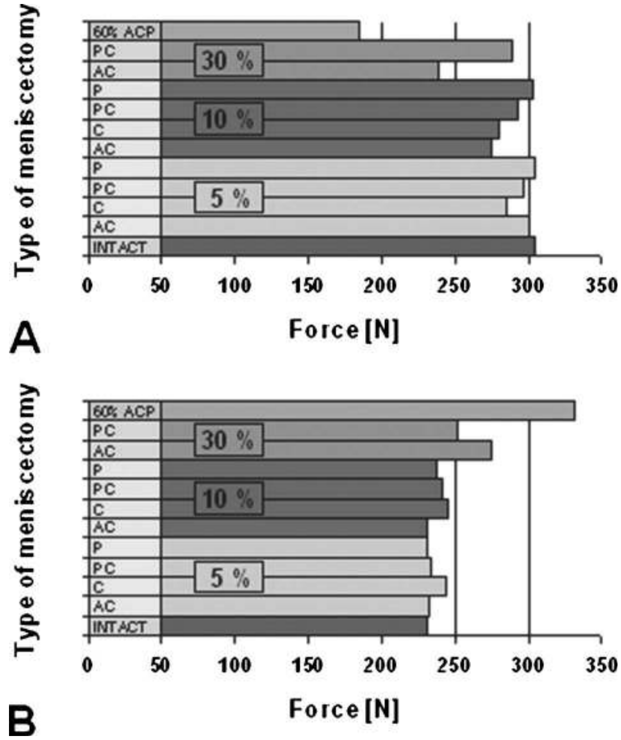

Fig. 8 (*A*) Contact force on the superior surface of the medial meniscus, (*B*) contact force on the tibia plateau—medial side

the medial meniscus strain, from approximately 9% to almost 18% (Fig. 7). The location of maximum axial strain for the articular cartilage on the medial tibial plateau is where the femur and tibia contact directly. However, on the lateral tibial plateau, the location of the maximum axial strain is under the central third of the lateral meniscus. The location of the maximum values of strain did not move following the various partial meniscectomies. Medial tibial plateau strains were approximately 15.5% for all cases, and increased to 16.7% when 60% of the medial meniscus was removed. The lateral portion of the tibial plateau experienced a maximum strain of 14.7% for the intact meniscus. The greatest increase was up to 15.6% strain when 30% anterior-central was removed.

In general, as the percentage of removed medial meniscal tissue increased, the total contact force decreased on the superior surface of the medial meniscus, with a corresponding increase in total contact force on the region of the tibial plateau in direct contact with the femoral condyles (Figs. 8(*A*) and 8(*B*)). No significant changes in total contact force were noted on the superior surface of lateral meniscus with increasing degrees of partial meniscectomy.

Discussion

The percentage of meniscus removed was proportional to contact pressure and inversely proportional to contact area for the superior surface of the medial meniscus. The location of meniscectomy affected the results since the distribution of contact pressure was not uniform throughout the meniscus. Removing tissue from regions of high pressure (posterior) did not change the contact area significantly, but resulted in large changes in the contact pressures. Conversely, removing tissue in regions of lower load bearing, such as anterior-central region, resulted in a more equal change in contact pressure and area. Such results are likely due to the fact, that the distribution of pressure on the anterior-central part of the meniscus spanned from the inner to outer edge of the meniscus, whereas contact in the posterior-central portion primarily occurred in the outer 1/3. The anterior and central portions were more sensitive to tissue removal than the posterior portion even though higher contact pressures exist in the posterior portion. It appears that removing only 5%–10% of the medial meniscus can increase pressures by as much as 20% in the anterior region and by as much as 141% in the central region. This analysis showed that the contact area on the superior surface of the medial meniscus is twice as small as other contact surfaces in the knee, likely contributing to the higher contact pressures on the superior surface and possibly suggesting a reason for a higher number of injuries to the medial meniscus.

Removing portions of meniscal tissue did not shift the location of maximum pressure or maximum strain on the superior surface of the medial meniscus. However, it is important to note that this study was only completed at 0 degrees of flexion and perhaps at other flexion angles meniscectomy may change the location of maximum pressures and strains. Brown et al., [28] showed a posterior shift in the overall contact pattern, but no statistical change in the contact area, mean contact stress, or in the maximum contact stress, with increasing flexion angle when the menisci were intact [28]. It is currently not known how a meniscectomy would affect this shift in pressure pattern. Possibly the ligaments support larger loads following meniscectomy in order to stabilize the joint. Future studies with this model will study how ligament tensions change following varying degrees of partial meniscectomy.

The current results show that the lateral meniscus is only slightly affected by medial meniscectomy. Changes of approximately 25% in the maximum pressure were seen when removing 60% of the medial meniscus, whereas changes never exceeded 7% in the contact area and mean pressure for any of the medial meniscectomies. Previous research agrees with these results, suggesting that the lateral hemijoint is not adversely affected by medial meniscectomy [43,44]. It appears that medial meniscectomy primarily affects the remaining medial meniscal tissue and the underlying cartilage. Increases as large as 55% on the medial tibial plateau maximum pressure were found following removal of 60% of the medial meniscus.

While the model provides a meaningful comparison of various partial meniscectomies, a few limitations are worth noting. The first limitation of this analysis is the simplified constitutive model used for the cartilage and meniscus. A single phase material model was used to represent both tissues. However, since we are only looking at contact pressures, under relatively short loading durations, this has been shown to be a valid model [26]. Secondly, we have only investigated one load level with the model, 1200 N, axial compression. It is likely that at larger flexion angles or higher loads the changes with meniscectomy may be more exaggerated. However, we note that the location of maximum pressure did not change with increasing loads from 0 to 1200 N. Another limitation of the model is that the geometry of only one knee was utilized to build the model. We expect that if the model were created with a different knee, having different geometry and material properties, the absolute numbers would be different, but the relative changes seen in this study would be similar for any healthy knee joint. The highly contoured surface of the superior meniscus and femoral condyles likely leads to an increase in the sensitivity of contact pressure to geometry. It is likely that our absolute values for contact were affected by joint geometry and the material properties of the articular cartilage, meniscus and ligaments. However, we would expect the relative results to hold true for a range of geometries and material properties.

The contact area measured in this model for the intact case is comparable with experimental results by Fukubayashi and Kurosawa [45]. The total tibial plateau contact area in the present study was 1,040 mm^2 and in Fukubayashi and Kurosawa's study they measured 1,150 mm^2 [45]. Comparing total contact area on the medial side, the present analysis resulted in 620 mm^2 whereas Fukubayashi and Kurosawa obtain 640 mm^2 [45]. Previous studies have found values for mean contact pressure on the superior (1.2 MPa) [28] and inferior (1.3 MPa) [13] intact meniscal surfaces. The present model found 1.57 MPa, for the intact superior meniscal surface, and 0.94 MPa for the inferior intact case. Brown et al., found a maximum contact pressure on the medial side of the femur condyles of 4.2 MPa; our analysis found 4.7 MPa [28]. Previous FE models have reported strains for individual nodes throughout the meniscal tissue. We reported the maximum change in height of the meniscus, which would be an average of nodal strains through the depth of the tissue. For comparison, if we just use the superior nodes of the meniscus, the maximum axial strain in this study was 13% for the intact case. The axial strain found in our model is similar to Spilker et al.'s 2D intact meniscus model where they obtain maximal strain of 14% [23].

In the future we will look at radial displacement in the meniscus which may be one of the precursors of OA. Spilker and Donzelli [23] found that increased meniscal radial displacement changed the load bearing capability of the menisci, and raised the stress concentration in the meniscus, which could lead to OA. Kenny et al. [46] found that the displacements are greater in a knee with Fairbank's signs (radiographic abnormalities in the knee after meniscectomy, which are a subset of the radiographic signs of OA [47]). However, some patients had large displacement occurring in the absence of symptoms. These data confirmed their hypothesis that large meniscal displacements precede Fairbank's signs. Therefore, noticeable changes in meniscal displacements could be the one of the first signs of OA.

Small changes in contact parameters following 5% and 10% meniscectomies were expected as small amount of tissue was removed. These results are promising for future research and can help surgeons to choose the best surgical treatment. However, more research has to be completed in this area to understand what load levels will contribute to tissue damage or OA. In fact it is unknown what changes in load cause irreversible changes in the tissue, possibly increasing catabolic activity of the cells, or matrix degradation. Future work will use data from the present analyses as input to a mechanical bioreactor to compress meniscal tissue and measure the biochemical response. Both intact and meniscectomized stresses and strains will be evaluated to better understand the biochemical milleau that ensues in the knee following partial meniscectomy.

Acknowledgment

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